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A Comparative Study of Electrical Potential Sensors and Ag/AgCl Electrodes for Characterising Spontaneous and Event Related EEG Signals

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Abstract

For exactly 90 years researchers have used electroencephalography (EEG) as a window into the activities of the brain. Even now its high temporal resolution coupled with relatively low cost compares favourably to other neuroimaging techniques such as magnetoencephalography (MEG) and functional magnetic resonance imaging (fMRI). For the majority of this time the standard electrodes used for non-invasive monitoring of electrical activities of the brain have been Ag/AgCl metal electrodes. Although these electrodes provide a reliable method for recording EEG they suffer from noise, such as offset potential drift, and usability issues, for example, difficult skin preparation and cross-coupling of adjacent electrodes. In order to tackle these issues a prototype Electric Potential Sensor (EPS) device based on an auto-zero operational amplifier has been developed and evaluated. The absence of 1/f noise in these devices makes them ideal for use with signal frequencies of ~10 Hz or less. The EPS is a novel active ultrahigh impedance capacitively coupled sensor. The active electrodes are designed to be physically and electrically robust and chemically and biochemically inert. They are electrically insulated (anodized) and scalable. A comprehensive study was undertaken to compare the results of neural signals recorded by the EPS with a standard commercial EEG system. These studies comprised measurements of both free running EEG and Event Related Potentials (ERPs). Results demonstrate that the EPS provides a promising alternative, with many added benefits compared to standard EEG sensors, including reduced setup time, elimination of sensor cross-coupling, lack of a ground electrode and distortion of electrical potentials encountered when using standard gel electrodes. Quantitatively, highly similar signals were observed between the EPS and EEG sensors for both free running and evoked brain activity with cross correlations of higher than 0.9 between the EPS and a standard benchmark EEG system. Future developments of EPS-based neuroimaging include the implementation of a whole head ultra-dense EPS array, and the mapping of distributions of scalp recorded electrical potentials remotely.

Keywords: EEG, Electrical Potential Sensor, Capacitive electrode, ERP, ud-EEG,

1. Introduction

Traditional methods of acquiring electroencephalogram (EEG) signals rely on the use of silver/silver chloride (Ag/AgCl) transducing electrodes. This type of electrode converts ionic current on the surface of the scalp to electronic current for amplification and subsequent signal processing. Ag/AgCl electrodes are cheap, and in clinical applications disposable, but require the use of a conducting gel between the electrode and the skin, since they rely on maintaining a low electrical resistance contact (Searle & Kirkup, 2000). When applying gel electrodes a low impedance path of $<10^4\ \Omega$ is usually achieved by abrading the scalp. This is then followed by an acquisition system with a typical input impedance of 10^6 to $10^7\ \Omega$. Although high, this impedance ratio nonetheless still causes distortion of the very surface potentials it is trying to measure. This may seem unavoidable: Since the electrical potentials that are the target for recording are caused by the flow of current in the brain, any device that requires a real charge current to flow through it in order to make a measurement, necessarily distorts the source of that potential.

Both in clinical and research fields the acquisition of clean EEG data requires highly skilled personnel. This is apparent especially during experimental setup. As well as the need for scalp abrasion (as mentioned), additional problems for standard EEG are that the conducting gel may cause skin irritation and discomfort, tends to dry out after a period of time, and needs to be washed out of the hair upon completion, meaning that these types of wet electrodes are unsuited for long term clinical monitoring applications (Prutchi & Norris, 2005). The gel may also lead to cross coupling or bridging between electrodes in an array if great care is not taken during placement, a problem which is exacerbated when using high-density EEG arrays.

Dry conducting electrodes, consisting of a benign metal (such as stainless steel) with no electrolyte between electrode and skin, provide a more user friendly approach, with the electrodes making resistive contact directly with the skin (Chi, Jung, & Cauwenberghs, 2010). This overcomes some of the problems caused by the wet gel electrodes, but introduces additional variables: the variation in contact resistance due to perspiration, skin creams etc. and their susceptibility to movement artefacts. For these reasons dry conducting electrodes tend to produce noisier signal measurements than wet electrodes (Chi et al., 2010). However, a clinical comparison of concurrent measurements with wet and dry EEG electrodes concludes that there is a high degree of correlation between the signals obtained from both types of electrodes. Some research suggests that dry conducting electrodes offer better long-term performance (Gargiulo et al., 2010).

An alternative approach is to dispense with the conventional resistive contact approach (either wet or dry electrodes) altogether, and instead to measure neuroelectrical activity *without* forming an electrical contact with the head. If an insulation layer is placed in front of an electrode, it provides a capacitive path between the sensor and the brain, which can provide an alternative readout of brain electrical responses. Such an approach is used in Electric Potential Sensors (EPS) which operate as high performance DC stable electrometers (Clippingdale, Prance, Clark, Prance, & Spiller, 1991). Using this method the signal fidelity no longer relies on skin resistance, which should have the effect of reducing the variation in the signal. This approach also dispenses with the signal distortion caused intrinsic to standard EEG electrodes, since the EPS does not require a real charge current to flow through it in order to make a measurement.

Electric Potential Sensor technology has already demonstrated its efficacy for research and clinical applications in electrocardiographic (ECG) data acquisition, where the inherent DC stability and short settling time of the EPS is advantageous compared to other insulated electrode implementations (Harland, Clark, & Prance, 2002; H. Prance, 2011). However, the low frequency noise performance required for accurate EEG data acquisition is considerably more stringent than for ECG, with average EEG signal strengths of 0.1 to 100 μV for EEG, as compared to ECG amplitude of 1 to 2 mV collected from electrodes placed on chest (Hampton, Adlam, & Hampton, 2008).

A review of sensor developments within healthcare settings discusses the low frequency noise performance of a number of active sensors and characterizes them in terms of the noise spectral density at 1 Hz (H. Prance, 2011). This can be used as a useful indicator of the performance of a new sensor for EEG purposes as it gives noise floor values ranging from 2 μV/√Hz to 10 μV/√Hz, which represents the intrinsic noise of the amplifier in the system; however these values will increase at lower frequencies due to 1/f noise.

The aims of the present study were: First, to assess the low frequency noise performance of high impedance capacitively coupled electrical potential sensors (EPS), to ascertain if it was analogous to, or lower than, conventional Ag/AgCl electrodes within a 0.1-10 Hz bandwidth. Secondly, to compare EPS and standard Ag/AgCl electrode EEG recordings from free running EEG and Event Related Potential (ERP) paradigms, in order to investigate if the resulting signals were broadly comparable.

Previously it has been demonstrated that the EPS are sensitive to fluctuations in free running brain activity, such as alpha (8-14 Hz) and beta oscillations (14-30 Hz), as well as 'alpha blocking', an increase in the prominence of the alpha rhythm when the eyes are closed that is replaced by a beta rhythm when the eyes are open (Harland et al., 2002). We sought to expand upon these findings by characterising alpha activity recorded simultaneously from EPS and a standard EEG system using Ag/AgCl electrodes, in terms of the raw signal, frequency spectrum and similarities between the two signals measured by cross correlation.

We then investigated if alterations in brain activity caused by stimulus-driven visual perceptual changes would, first, be distinguishable by the EPS, and, secondly, display a similar evoked profile to standard EEG measures. Event-related potentials (ERPs) provide a good test of the sensitivity of the EPS since the maximum amplitudes of ERPs are small compared to ongoing background EEG activity, ranging from less than a microvolt to just several microvolts, compared to amplitudes in a typical adult raw EEG signal of between 10–100 μV. To investigate these questions we used a standard clinical paradigm known to elicit a robust ERP, the visual evoked potential (VEP) response, which is commonly used to assess pathology (and normal function) of the visual system (Odom et al., 2010).

We also investigated if the EPS would also be sensitive to changes in the category of visual stimuli presented by using a classic face perception paradigm known to reliably elicit early visual processing ERP components (Calder, 2011). Early ERPs that have been previously associated with face processing include the P1 and the N170. These are assumed to reflect, respectively, the extraction of fine/local information from a stimulus (Herrmann, Ehlis, Ellgring, & Fallgatter, 2005) and face-specific structural encoding (Eimer, 2000). When compared to other stimulus categories, face stimuli consistently elicit a larger negative deflection in the ERP from around 150-200ms (N170) after stimulus onset over occipitotemporal electrodes. Typically the N170 elicited by inverted face stimuli displays larger amplitudes over the right hemisphere and also occurs at a later time point compared to face stimuli of normal orientation (Eimer, 2000).

In summary, the present set of experiments sought to examine the sensitivity of the EPS to neuro-electrical responses, and to compare performance with standard EEG, at three levels of granularity: sensitivity to background activity, to event-related responses, and to stimulus category-specific responses.

2. Methods

2.1 Prototype Sensor and System

The prototype Sussex EPS is based on an auto-zero operational amplifier, chosen to give the minimum low frequency noise. The absence of 1/f noise in these devices makes them ideal for use with signal frequencies ~10 Hz or less, with a quoted noise performance of 22 nV/√Hz and 5 fA/√Hz. The input capacitance is ~8 pF with an associated voltage noise between 0.1-10 Hz of 0.5 μVp-p. After consideration of the expected signal amplitudes and frequencies the sensor was configured to have an operational bandwidth of 0.1 Hz to 78 Hz and a voltage gain of x50. The voltage gain was distributed between two stages with x5 and x10 respectively for the first and second stages (for details of the operation of EPS devices see (Harland et al., 2002), for block diagram see Figure 1).

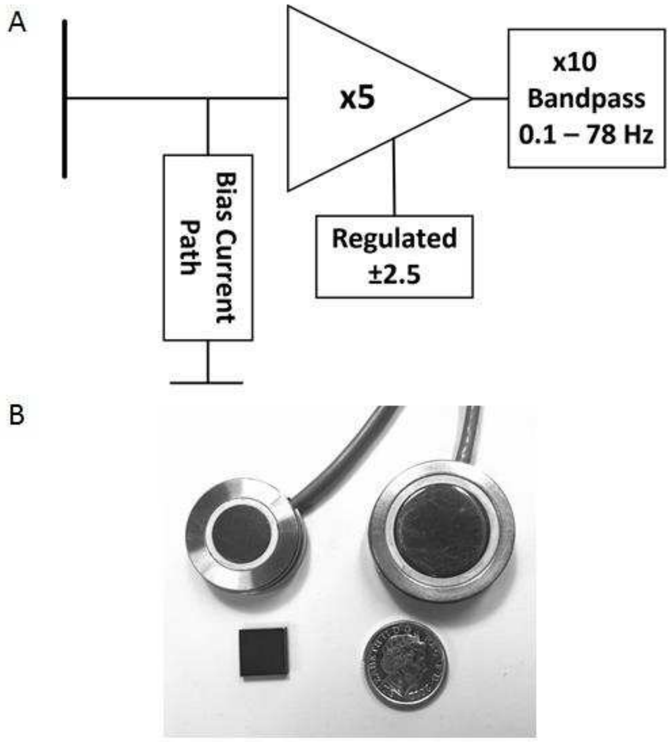

Figure 1. A. Block diagram of an Electrical Potential Sensor. B. Image of the two sizes of Electrical Potential Sensor used in this study (12 mm and 18 mm), the microchip version of the Electrical Potential Sensor for ECG purposes (not used in this study) and a 5 pence coin, shown for scale.

The sensors were operated from split symmetric power supply rails of ±2.5V. Two different electrode size versions were used in this study, either 12 mm or 18 mm diameter, to enable reliable contact to be made to different locations on the scalp and mastoids. The electrodes are electrically insulated through an anodized layer. Both sensors were housed in inert stainless steel machined housings with the electronics fabricated in surface mount on a PCB compatible with epoxy potting compounds. The sensors are designed to be immersed in alcohol for sterilization purposes.

The gain and operational bandwidth of the sensors was confirmed using a standard spectrum analyzer. The most significant parameter for the specification of the sensor in this particular application is the voltage noise referred to the input. This was measured by placing the sensor in a screened environment and recording the spectral noise density over a 1 kHz bandwidth. From this data, shown in Figure 2, two values are produced to characterize the noise performance, the spot noise value at 1 Hz and the integrated noise from 0.1 Hz to 10 Hz. The results obtained for the voltage noise measurements are: 30 nV/√Hz at 1 Hz and 0.2 μVp-p from 0.1 to 10 Hz; consistent with the data provided by the manufacturer (Analog Devices Inc, USA). The absence of 1/f noise in this data confirms that the auto-zero amplifier used in this study was performing as expected.

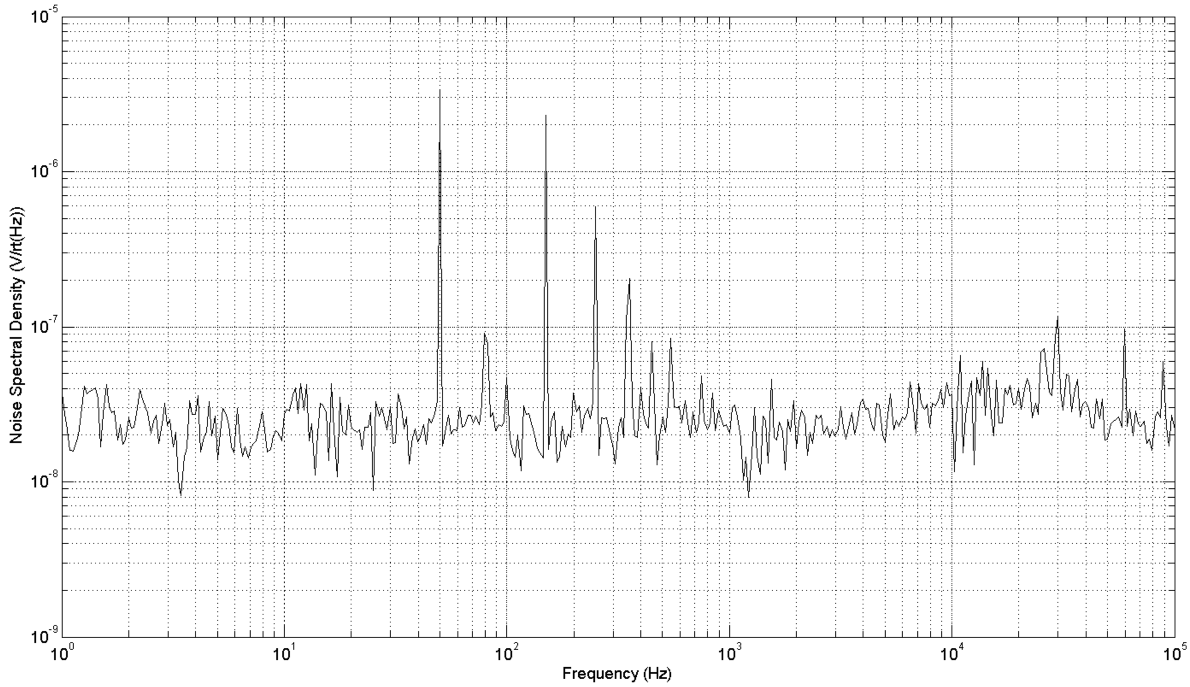

Figure 2. Noise spectral density plot for prototype EPS with auto-zero amplifier eliminating the 1/f noise of typical input stage of acquisition system.

In order to confirm, at an early stage in the design process, that the EPS design was both suitable for high quality EEG signal acquisition and that it was compatible with commercial systems and practice, we interfaced the sensors with a 64 channel Refa 8 amplifier (ANT Neuro, The Netherlands) via bipolar ExG inputs at 24 bit resolution with an input noise of 1 μV_{rms}. This enabled us to make direct comparisons between the EPS and wet gel electrode measurements using the same amplifier. All electrode cables had active shielding to reduce 50 Hz mains interference and cable movement artefacts.

3. Experimental Methods

Both types of sensors were placed accoding to the International 10-20 system. All data was recorded using a 64 channel ANT Neuro amplifier (Refa 8) at a sampling rate of 2048 Hz. The Refa amplifier has an internal gain of 26.55 which is removed by the ASAlab software, and the EPS also had an inbuilt gain of 50 which we compensated for in post-processing.

3.1 Experiment 1: Alpha Characterisation

3.1.1 Methods

Three participants took part in this experiment. For participants 1 and 2, data were collected in two separate blocks, one using a 64 channel Waveguard EEG cap (ANT Neuro, Enschede) employing standard Ag/AgCl electrodes, and another using the EPS. EEG (Waveguard) was measured from electrode O1, plus left and right mastoids, and online re-referenced to linked mastoids. EPS sensors were placed at similar positions, and re-referenced in the same way. Participants were seated in a dimly lit electromagnetically shielded room and asked to relax and stare straight ahead. Participants were asked to alternately close and open their eyes every time they heard an auditory signal, which occurred approximately every 4 seconds. Recording lasted approximately 60 seconds for each block. Data was offline detrended and filtered using a bandpass filter between 0.1 and 80 Hz.

For participant 3, (Ag/Cl) EEG and EPS data were measured simultaneously. EEG was measured from electrode Oz and online re-referenced to electrode Fz. EPS electrodes were placed under the EEG cap between Oz and O1 (Oz-EPS) and between Fz and F1 (Fz-EPS), with the Oz-EPS online re-referenced to Fz-EPS. The recording conditions and task were the same as for the other two participants, but overall recording duration was two minutes. This modification was made to allow a direct comparison between the two systems using simultaneous recording. Initial observation of the data for all 3 subjects showed that ocular muscle artefacts caused by opening, closing or blinking the eyes led to small amounts of drift in the signal measured at Oz by both systems. Interestingly, the two systems reflected this drift to different degrees such that whilst the underlying higher-frequency signals were very similar, momentary lower-frequency differences in the range of up to 2 Hz created larger-scale drifts between the two. Because these low frequency drifts are *generally* considered to be less important, and are usually eliminated in event-related analyses by baselining techniques (Luck, 2005), we chose to remove these trends by offline filtering the EEG and EPS data from participant 3 between 2 and 80 Hz. Future studies may focus on the possible causes and interpretations of these differential low-frequency drifts.

3.1.2 Results

Figure 3 shows a representative sample of EPS and EEG data from participants 1 and 2 during 4-second periods when they transitioned from eyes-open to eyes-closed. Both participants clearly show the characteristic increase in alpha activity (~10 Hz) when they closed their eyes (at approximately the 2 second mark), and this type of activity is apparent in both EPS and EEG recordings. Figure 4 shows power spectral density, measured by Welch's method (Welch, 1967), over the entire 60 second recording period. As can be seen, the spectral density patterns are very similar between the ANT EEG and the EPS sensors, indicating that the EPS measures similar underlying neural activity to the standard EEG electrodes.

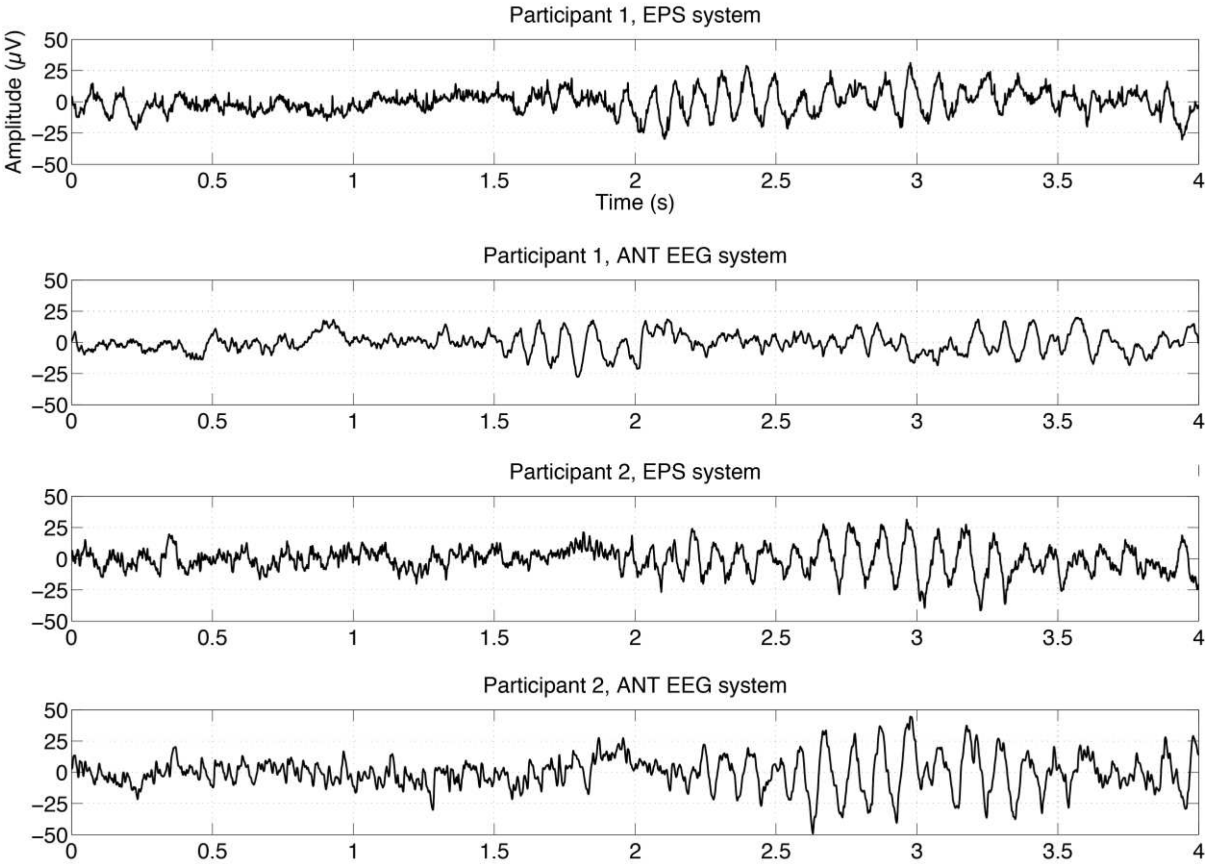

Figure 3. Representative raw data recordings from Participants 1 and 2, from both ANT-EEG and EPS systems, over 4 second periods including transitions from eyes open to closed (2 s). Note patterns of alpha band activity after eyes are closed, which appear similar for both the ANT-EEG and EPS systems.

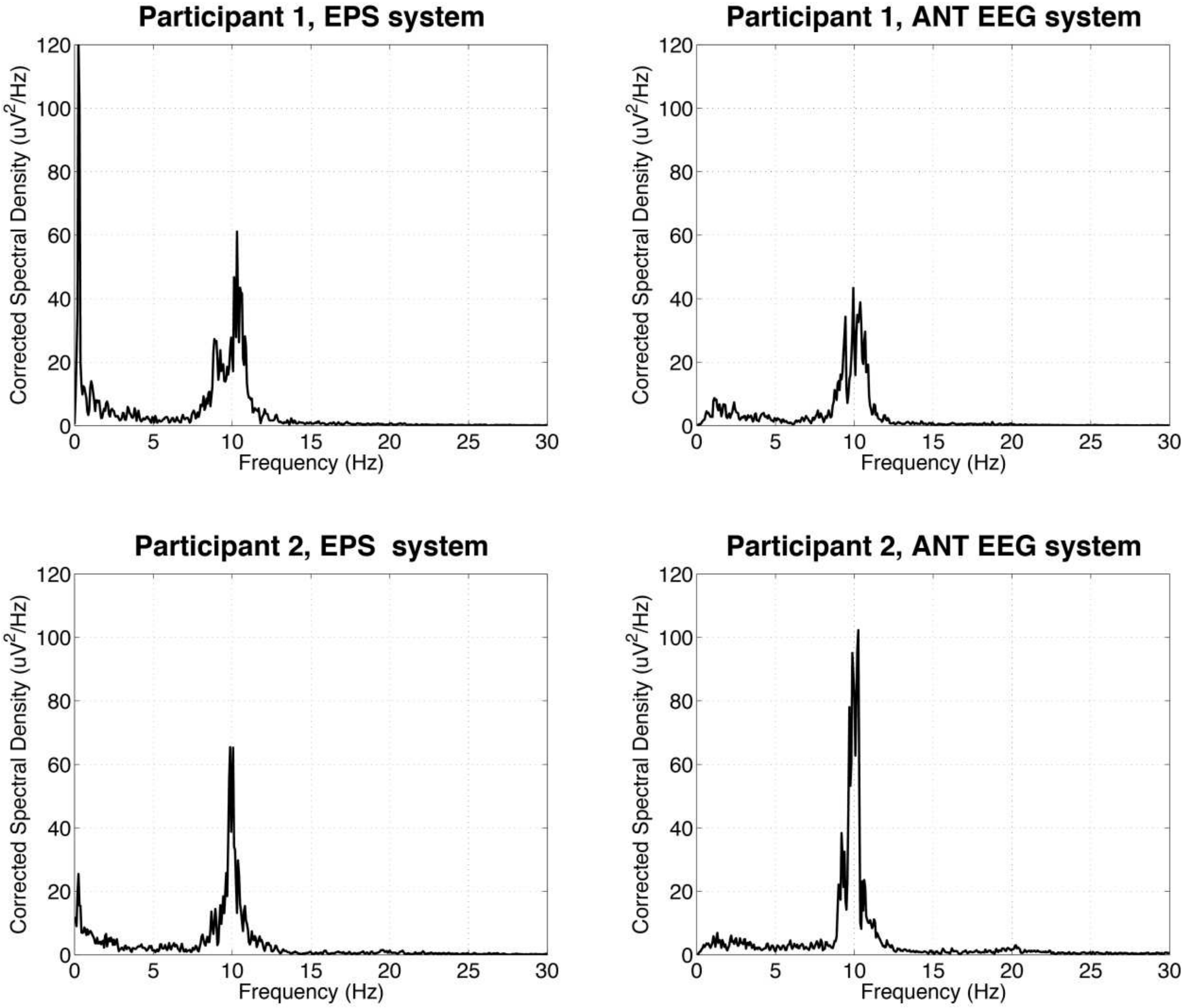

Figure 4. Frequency spectral density plots from the full 60 s recordings for Participants 1 & 2, for both ANT-EEG and EPS systems.

Figure 5 displays results from participant 3, where data was collected from EEG and EPS simultaneously and from as-close-as-possible electrode locations. Figure 5a shows six seconds of the same data recorded from electrode Oz during a transition from eyes open to eyes closed (at around the 1 second mark). Note the remarkable similarity in activity measured by the two systems. Figure 5b shows the frequency spectra over the two minutes' worth of recorded data from the EEG and EPS systems at the Oz location. The spectra are virtually identical. Figure 5c shows a cross-correlation between the data from the two systems over the entire 2 minute recording session. Cross-correlation value is highest around lag zero, sharply reducing to around r = 0 at either side. These data provide compelling evidence that the signals measured by the two systems are near identical, at least in this setting, strongly indicating that they are both measuring the same underlying neural activity.

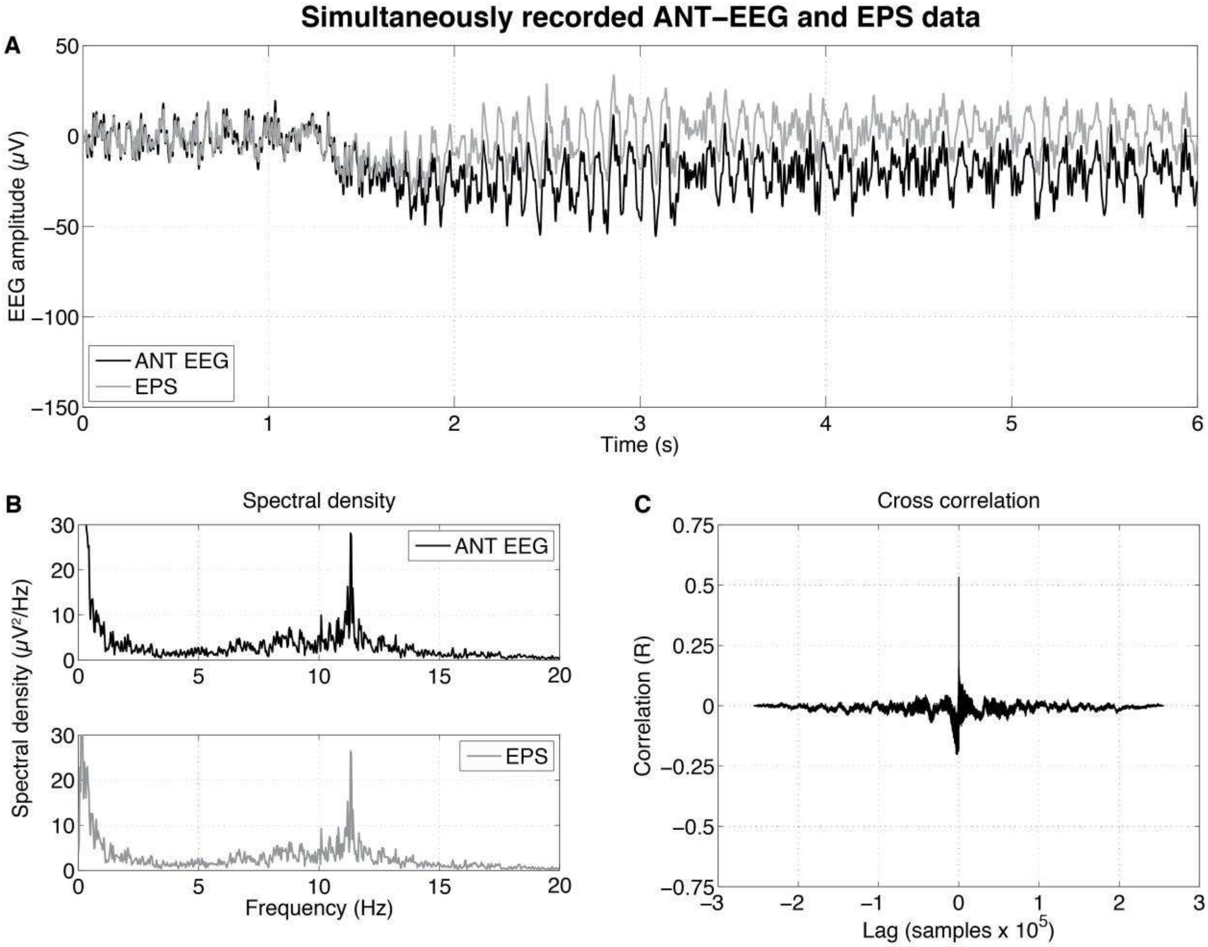

Figure 5. A: 6 seconds of simultaneously recorded EEG and EPS data during a transition from eyes-open to eyes-closed. B: Frequency spectra for EEG and EPS data over a 2 minute period. C: Cross-correlation between the two systems over a two minute period.

3.2 Experiment 2: Visual Evoked Potentials

3.2.1 Methods

A single participant was seated in a dimly lit electromagnetically shielded booth, 75 cm away from a LaCie Electron blue IV 22" CRT Monitor set at 1024 x 768 pixel spatial resolution and a refresh rate of 100 Hz. Stimuli consisted of a high-contrast checkerboard of black and white checks that changed phase (i.e., black to white and white to black) abruptly and repeatedly at two reversals per second (1Hz)(Pattern Reversal). Inter-stimulus interval was jittered between 200 and 600 ms in 50 ms steps. The design of the experiment was based on the International Society for Clinical Electrophysiology of Vision (ISCEV) guidelines for eliciting visual evoked potentials (Odom et al., 2010). The luminance and contrast of the stimulus was uniform between the centre and the periphery of the field. All stimuli subtended 16° of visual angle and were presented using Matlab 2013a

(Mathworks, USA) and the Psychophysics Toolbox ((Brainard, 1997; Pelli, 1997); www.psychtoolbox.org). A small circular red fixation dot at the centre of the visual field was present in both experiments to control central fixation and maintain visual attention. Central fixation was continuously monitored throughout the experiment via observation of the vertical and horizontal eye channels. Block 1 was a practice and comprised of only 5 stimulus reversals of the checkerboard. The experimenter was in the same room during the practice to check whether participant understood the instructions and carried out the task appropriately. For the remaining seven experimental blocks, participants were alone in the room. Each block included 100 trials. Blocks 2 and 3 involved the presentations of checkerboards at two reversals per second (1Hz) (array of 16 x 16 checks of 1° per side of each square). Between blocks participants were allowed to take a break, move, stretch or ask any questions to the experimenter. The non-dominant eye of the participant was covered using an eye-patch and their head was supported on a chin rest so that their line of sight was exactly central to the screen.

EEG and EPS data were measured simultaneously. EEG was recorded using a 64 channel ANT Waveguard cap from Oz and Fz. Data were online re-referenced to electrode Fz (Odom et al., 2010). EPS electrodes were placed under the EEG cap between Oz and O1 (EPS-O1) and between Fz and F1 (EPS-Fz) and connected to the amplifier via the bipolar ExG inputs. EPS signals were measured from EPS-O1 and EPS-Fz and additionally a differential of EPS-Fz and EPS-O1 was also recorded. Two additional Ag/AgCl electrodes were also used to record vertical ocular artefacts.

EEG data was analysed offline with EEGLAB (Delorme & Makeig, 2004) and ERPLAB Toolbox (Lopez-Calderon & Luck, 2014). Data was band-pass filtered from 0.1 to 30 Hz with a Butterworth digital filter and target locked epochs were created. Each epoch started 100 ms before the onset of the target and ended 300 ms afterwards. Ocular artefacts were identified using a moving window peak-to-peak threshold, with a voltage threshold of 40 μV. Epochs were then baseline corrected for -100 to 0 ms before stimulus onset and then averaged to produce VEPs.

3.2.2 Results

As can be seen from Figure 6a VEPs for both EPS and ANT EEG sensors display a very similar pattern of evoked responses. Both display standard evoked responses for pattern reversal stimuli including the N75 and P1 VEP components (Odom et al., 2010).

Cross-correlation value is highest around lag zero, sharply reducing to around $r = 0$ at either side. Apart from the evident visual similarities in the VEPs recorded by the EPS and EEG sensors the results of the cross-correlation provides additional evidence that the signals measured by the two systems are similar, and that they are both measuring the same underlying visually evoked neural activity.

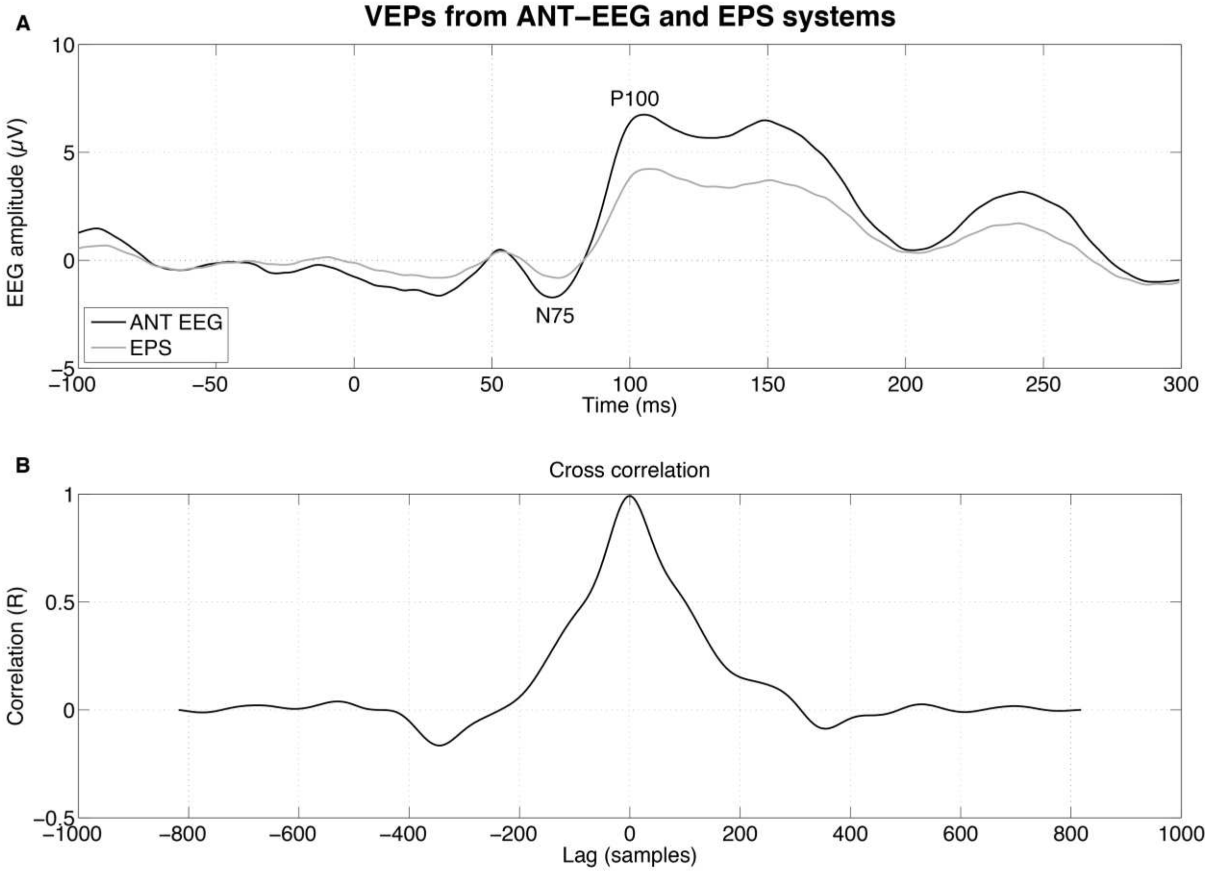

Figure 6. A: Averaged VEPs from pattern reversal for EEG (black line) and EPS (grey line). B: Cross-correlation between the two systems from -100 – 300ms.

3.3 Experiment 3: Face Processing

3.3.1 Methods

Four subjects took part in this experiment. Each sat 75 cm away from a LaCie Electron blue IV 22" CRT Monitor set at 1024 x 768 pixel spatial resolution with a refresh rate of 100 Hz. For each trial subjects were presented with a central fixation cross, the duration of which was randomly jittered in steps of 40 ms between 500-900ms. Subjects were then randomly shown one of 3 categories of visual stimuli: faces, inverted faces (taken from Utrecht ECVP) and scrambled face images (Figure 7), which were each presented at the centre of the screen for 350 ms. All stimuli occupied a visual angle of approximately 5.5° x 4°. Subjects were then shown a blank screen lasting 800ms, followed by a '?', which was presented for 1500ms, during which subjects had been instructed to press the space bar, using their dominant hand, if the stimulus presented was an upright face. Stimuli were presented using E-Prime 1.2 software (Psychology Software Tools, Pittsburgh, PA, USA). All stimuli were matched for luminance and displayed in frontal view. Scrambled faces were created by randomly rearranging the pixels of upright face stimuli. There were a total of 360 trials, 120 stimuli were presented in a random order for each participant from each of the three categories. Participants were instructed to maintain central eye fixation during the experiment.

Figure 7. Examples of stimuli used in this study: From left to right, Face, Face inverted, Scrambled Face.

Subjects performed the experiment twice, first with the EPS and then with EEG sensors. Counterbalancing the order was not possible, as participants' hair needs to be washed after EEG recording to remove gel, which would render the hair too damp (within a reasonable experiment time) to subsequently measure EPS. Three EPS were used placed at M1, M2 (12 mm) and P8 (18 mm) and were was recorded at a sampling frequency of 2048 Hz, with the left mastoid as a reference. Two VEOG and two HEOG AgCl electrodes were used to record ocular artefacts for both EEG and EPS recordings. EEG was recorded with a Waveguard 64-channel cap, also with the left mastoid as a reference. To maintain consistency with the EPS setup three electrodes were used: M1, M2 and P8. All EEG/EOG electrode impedances were kept below 5 KΩ.

EEG data were analysed offline with EEGLAB (Delorme & Makeig, 2004) and ERPLAB Toolbox (Lopez-Calderon & Luck, 2014). Data was band-pass filtered from 0.1 to 30 Hz with a Butterworth digital filter and target locked epochs were created for each stimulus category, Face, Face Inverted and Scrambled Faces. Each epoch started 200 ms before onset of the target and ended 500 ms afterwards. Ocular artefacts were identified using a moving window peak-to-peak threshold, with a voltage threshold of 40 μV. Epochs that exhibited excessive noise (+/- 100 μV) were manually rejected. This rejection criteria led to an average of 10 Face, 8 Inverted and 10 Scrambled trials being discarded from the ANT recordings and 23 Face, 17 Inverted and 16 Scrambled trials for the EPS. Epochs were then baseline corrected for -200 to 0 ms before stimulus onset. Stimulus locked epochs were then created for correct trials for the three conditions Face, Inverted Faces and scrambled faces and then averaged across each condition.

3.3.2 Results

Figure 8 shows that the EEG and EPS recordings produced remarkably similar early ERPs and standard evoked responses to face stimuli. Critically, the early ERPs associated with face processing - the P1 and N170 components – are easily observable and very similar in the EPS and EEG data. We also replicated, for both sensor types, the finding of a pronounced N170 over the right hemisphere that occurs at a later time point for inverted face stimuli as compared to (non-inverted) face stimuli (Eimer, 2000). The results demonstrate that the EPS are sensitive to changes in neural activity associated with differing categories of stimuli, and are sensitive to face-specific ERP responses.

Another factor that was also examined in this experiment was the EPS sensitivity to movement artefacts, as subjects were asked to signal with a button press the onset of upright faces. While the subjects response occurred at a later point to the time window shown in Figure 8, previous research comparing insulated capacitive sensors has shown both an increased sensitivity to motion artefacts and extended settling time compared to standard EEG sensors (H. Prance, Watson, Prance, & Beardsmore-Rust, 2012). However, as can be seen from Figure 8 there is no notable alteration of the EPS ERP components for upright Faces when comparing the EPS response to other categories of stimuli.

Overall, these findings confirm that the EPS displays a sensitivity profile that can detect category specific changes in amplitude of ERPs within the background of ongoing EEG activity.

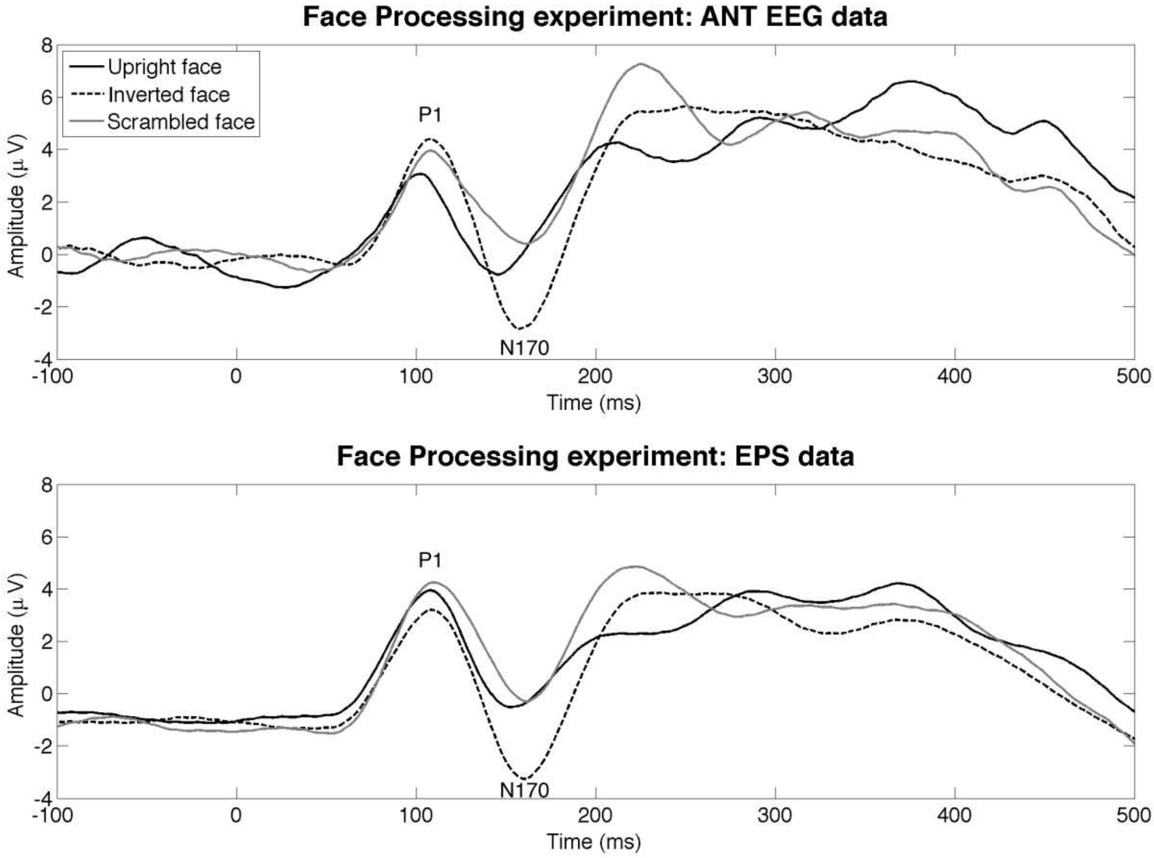

Figure 8. Grand-average ERPs (n=4) displayed between -100 and 500 ms for 3 categories, upright faces, inverted faces and scrambled faces over occipitotemporal electrode (P8).

4. Discussion

Our study aimed to compare the signal and signal-to-noise properties of our novel EPS sensors to an optimized conventional wet gel Ag/AgCl electrode EEG system. The results demonstrated that EPS can readily measure both free running EEG and event related potentials (ERPs), producing a strictly comparable signal and signal to noise ratio to conventional wet gel electrodes. In addition, the low frequency noise profile of the EPS is highly suited for accurate EEG data acquisition. Experiment 1 showed that ongoing oscillatory activity is detected equally sensitively by the EPS prototype and by the EEG system, where measurements of alpha activity were remarkably consistent between the two systems, when modulated by opening and closing the eyes. Further experiments showed that the EPS system is suitable for measuring event-related averaged components that correlate with both early-sensory (Experiment 2) and perceptual-cognitive processes (Experiment 3). The Sussex EPS prototype was designed as a proof of concept, and we now focus on the benefits of the EPS compared to conventional sensors and some of the potential future applications of these sensors.

4.1 Benefits and Drawbacks of EPS

The EPS system, even at prototype stage, has many advantages over conventional EEG sensors. These include reduced setup time, elimination of sensor cross-coupling, no requirement for a ground electrode, and absence of distortion of electrical potentials encountered when using standard gel electrodes. When using conventional gel electrodes a connection between the scalp and metal conductor is achieved through the application of an electrolyte solution. The electrical properties of this interface, known as a half-cell potential, are governed by the electrochemical reactions between the two layers (Chi, 2010). The electrochemical reactions produce fluctuations in the metal-electrolyte potential that can cause an increase in noise levels of up to 10uV peak-to-peak for Ag/AgCl electrodes (Geddes, 1972). Unlike conventional electrodes, the EPS has an insulation layer that is placed in front of the electrode, meaning that there is no physical contact between the metal and an electrolyte (such as sweat or gel) and as a result half-cell potentials are not an issue with this method of measuring brain activity.

The EPS does display a higher sensitivity to motion artefacts, both of the subject and surrounding environment compared to standard EEG recordings, this is due to the effect of motion within the local environment on the surrounding electrical fields (H. Prance et al., 2012). However, within the context of EEG recordings participants are generally instructed to remain as still as possible during the experiment and are also usually located within a separate recording chamber, which minimises the influence of these artefacts on data recorded using the EPS. Searle et al., (2000) found that both Dry and Insulating electrodes are more susceptible to movement artefacts than wet electrodes. However, after the Insulating electrodes were allowed to settle, they showed lower levels of movement artefacts compared to standard wet electrodes. Indeed we found in all three experiments that once the EPS were applied to the scalp the influence of motion artefacts was no greater than with the standard EEG system.

4.2 Future Directions

While the current prototype EPS sensors are rather large, precluding deployment of high-density arrays in EEG settings, we are actively pursuing minimization of EPS (this has already been achieved in the context of ECG, which has seen the sensors miniaturized to the microchip level 10mm x 10mm, see Figure 1). Future investigations of EPS/EEG will focus on the merits of using high (hd - 128-256 electrodes) and ultrahigh-density (ud >256 electrodes) montages to better represent the encephalogram (Odabaee, Freeman, Colditz, Ramon, & Vanhatalo, 2013; Petrov et al., 2014). The standard 10/20 placement system currently involves the placement of 21 electrodes approximately 6 cm apart.

Within the last 20 years the popularity of high density EEG (128-256 electrodes) has increased (Tucker, 1993) with the aim of localizing sources that drive the scalp recorded EEG signals. While the problem of volume conductance has been widely cited as a critical issue obstructing this goal, it is nonetheless clear that scalp recorded EEG signal does exhibit distinct spatial characteristics that underpin useful inferences about cortical sources. Until recently the available informational content extracted from scalp recorded EEG, referred to as “spatial patterning”, was thought to be quite limited. This is due to the assumption that there is a larger degree of smearing of source potentials by the volume conducting medium, especially the poorly conducting skull, distorting the spatial information recorded by scalp electrodes. However, there is evidence to suggest that by increasing the number of electrodes covering the scalp there is additional spatial information available that has previously been overlooked (Odabaee et al., (2013; Petrov et al., 2014). Investigations that have attempted to calculate the necessary inter-electrode spacing required to fully capture all the spatial information and spectral density available from scalp recorded EEG has resulted in estimates of around 2-3 cm (Srinivasan, 2005). However, Odabaee et al., (2013) found that sensor distances of between 5-10 mm are required to capture the full spatial texture of the raw EEG signal on a neonatal scalp. Petrov et al., (2014) found that even when using ud-EEG with a relatively small inter-electrode spacing of 1cm, there were strong variations in the EEG signal, measured using VEPs. Further, the use of this array led to a two-fold increase in the signal to noise ratio compared to a standard hd-EEG system. So it seems that an inter-electrode spacing of between 0.5-1cm is necessary not only to capture the full spatial texture of scalp recorded EEG, but also to increase the signal to noise ratio. However the technical challenge of deploying such a ud-array over the entire scalp using standard Ag/AgCl systems has meant that to date only a small ud-array of electrodes placed over a specific region has been achievable (e.g. 16 electrodes Petrov et al., (2014)).

Future miniaturisation of the Sussex EPS sensors would allow the creation of ud-EEG montages that could conceivably cover the entire head. The ability of miniaturised EPS sensors to couple capacitively to the scalp removes some of the major issues facing standard ud-EEG systems, such as the elimination of cross-coupling between electrodes caused by the use of electrolyte or gel on electrodes in close proximity, and the extensive set-up time that would be required for similar ud-standard EEG system (Merletti, 2010). Using a microchip version of the EPS sensor with dimensions of 10mm x 10mm, the resulting ud-montage could be as extensive as >1000 electrodes, surpassing theoretical (but not practically realised) ud-EEG setups of up to 800 electrodes (Petrov et al., 2014).

The relatively low input impendences of conventional EEG acquisition systems that use AgCl electrodes (10^6-$10^7\,\Omega$) give rise to the problem of volume conduction, which has been shown to significantly distort scalp recorded electrical potentials (Blum & Rutkove, 2007). Volume conductance is a general property of electrical currents to follow the path of least resistance, which causes activity from a cortical dipole to spread out further from the source and also to be smeared or diverted as it tries to pass through the highly resistant skull (Harland et al., 2002). These limitations have been addressed in other imaging techniques by using SQUID (super conducting quantum interference device) magnetometer systems in magnetoencephalography (MEG), which are less susceptible to volume conductance compared to Ag/AgCl electrodes and which do not require direct contact with the scalp (Ahonen et al., 1991). Unfortunately, SQUID systems are very expensive, mainly due to the cryogenic cooling of the sensors and the necessity for a magnetically shielded chamber to attenuate the Earth's own large magnetic field. However, it is clear from the data published using SQUID magnetometers that the recording of signals with no electrical connection to the body affords great benefits, including much improved reconstruction of the underlying sources of scalp recorded signals (Andreassi, 2007). The combination of a high level of sensitivity and very high impedances found in the EPS allows the accurate measurement of electrical potentials from the brain, within a minimally shielded environment. Additionally, and similarly to SQUID systems, the EPS does not require a ground electrode. This enables a direct comparison of activity between two different areas of the brain.

Finally, it has been shown that it is possible to remotely record ECG from up to 1m away from a subject using the EPS (Harland et al., 2002). Future studies investigating the EPS will focus on the possibility for remote detection of EEG, in a similar (more cost effective) manner to the SQUID magnetometers, to assess if a similar increase in the spatial resolution is gained with the EPS.

5. Conclusion

We have examined whether a novel electrical sensor, the EPS, could provide an innovative means for measuring electrical brain activity in experimental situations. Our data clearly demonstrate that the EPS provides a suitable alternative, with many added benefits, to standard EEG sensors. The EPS fulfils all of the necessary criteria of a sensor for recording scalp electrical potentials: it draws no current from the scalp and thus does not disturb the signal source, making it safe to use; it has an ultra-high impedance; it has a high tolerance to noise; it is straightforward to apply to human subjects; and it has a relatively low cost. Critically, even our prototype EPS/EEG shows similar sensitivity and signal-to-noise performance when measuring both ongoing neural oscillations and transient ERPs.

The EPS sensors intrinsic lack of cross-coupling and fast set-up time makes the implementation of a whole head ud-EEG array using the EPS an achievable goal. Such a technology could significantly influence neuroimaging research, allowing researchers to investigate more accurately the spatial dynamics of scalp recorded EEG, augmented by a higher SNR as well as applicability to a broad range of clinical settings. Additionally, mapping the distribution of scalp recorded electrical potentials remotely, in conjunction with ud-EEG using EPS, theoretically has the potential to resolve either a partial or full reconstruction of the location of their neural sources and therefore assist in unravelling the inverse problem.

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